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Simulations of Head Strikes in Helicopters and the Roles of Restraints, Seat Stroke and Airbags on Their Reduction (Reprint)

By

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SIMULATIONS OF HEAD STRIKES IN HELICOPTERS AND THE ROLES OF RESTRAINTS, SEAT STROKE AND AIRBAGS ON THEIR REDUCTION

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1. SUMMARY

Injuries from head strikes remain the leading cause of fatalities in U.S. Army helicopter mishaps. The roles of the restraint system, energy absorbing seat stroke and airbags in preventing or reducing the severity of head strikes are explored in this paper using mathematical simulations. Starting with a baseline simulation of an actual AH-64 survivable mishap in which the pilot received fatal basilar skull injury, the effects of three parameters were examined: timing of inertia reel locking, stroking of the energy absorbing seat, and the presence of an airbag mounted at the instrument panel. Results of the simulations suggested that delay of inertia reel in locking at the appropriate time together with obstruction of seat stroking may have caused the pilot's head to strike the glare shield. When a head strike was unavoidable, simulations indicated that an airbag would have reduced its severity.

2. INTRODUCTION

The crashworthy design of modern Army helicopters has resulted in fewer injuries from the impact acceleration in survivable crashes. The injury reduction, primarily to the spinal column, may be attributed to the energy-absorbing seat design which limits the forces transmitted to the seated pilot. Head and upper torso injuries also have been addressed with various design concepts to cockpit interior components, such as the breakaway optical relay tube used by the gunner in the AH-64 Apache helicopter. Following the introduction of these energy-absorbing devices into the U.S. Army Apache and Black Hawk helicopters, the injuries sustained in Army helicopter crashes between 1980 and 1985 due to excessive accelerations have dropped relative to other helicopters [1]. Ten years later, the risk of injury to U.S. Army helicopters occupants during the 1990-94 period was reduced significantly, primarily due to a 50% drop in head injuries [2]. Despite the success of the crashworthy design of modern helicopters, flail injuries continue to occur and, in fact, outnumber acceleration injuries. Contact or flail injuries are produced in secondary collisions which result from inadequate restraints, collapsing structure, or a combination of both.

Total delethalization of U.S. Army helicopter interior systems is impossible because of operational requirements and design constraints. Further, current restraints systems are unable to prevent secondary impacts [3]. The use of some airbag protection for the gunner has been suggested for many years [4], but no acceptable system ever was introduced into Army helicopters. More recently, the U.S. Army Aeromedical Research Laboratory (USAARL), Fort Rucker, Alabama, has demonstrated the effectiveness of airbags in reducing the severity of head injury [5], and evaluated the projected effectiveness of airbag supplemental restraint systems in Army helicopters [6]. These studies and other factors convinced the Army of the need to introduce airbag technology as a method of delethalizing the cockpit interior of its helicopters. As part of a development program by the Aviation Applied Technology Directorate to reduce the likelihood that aviators will be injured seriously by cockpit strikes [7, 8], Simula, Inc. developed a multi-airbag system [9] which will inflate during a crash to protect the aviator.

2.1 Objectives

In this paper, the roles of the inertia reel, energy absorbing seat, and airbags in preventing or reducing the severity of head strikes in helicopter crashes are explored by performing biodynamic simulations of crashes under various crash scenarios.

2.2 Baseline scenario

The simulations revolve around a baseline scenario in which a U.S. Army Apache helicopter crashed during a training mission at Fort Rucker, Alabama. The mishap resulted in fatal injuries to the rear seat pilot and survivable injuries to the front seat co-pilot. The seating configuration of the two pilots within an Apache is shown in Figure 1. An accident investigation team from the U.S. Army Safety Center gathered data of damage to the aircraft and cockpit, and medical assessment of the injuries sustained by the two pilots were made. The helmets, inertia reels, restraint harnesses, and crashworthy seats also were retrieved and examined by investigators from U.S. Army Aeromedical Research Laboratory to assess whether these life saving equipment functioned as expected. This allowed the investigators to infer kinematics of the aircraft prior to ground impact and to estimate the motion of the restrained pilots during the mishap. It was theorized that excessive extension of the shoulder harness may have been due to delay of the inertia reel in locking early in the mishap. The energy-absorbing seat stroked only about 2.5 cm due to distortion in the cockpit structure. These factors may have contributed to the head strike of the pilot with the instrument panel-mounted glare shield. The positions of the glare shield relative the pilot's head are shown in Figure 2 prior to and after impact with the ground.

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3. BIODYNAMIC SIMULATIONS

A widely used tool for accident reconstruction is the articulated total body (ATB) simulation software [10, 11]. Given a number of body segments connected by mathematical models at common joints, the ATB automatically formulates the differential equations that govern the motion of the body segments.

The model is driven by acceleration pulses which approximate the crash profiles. The ATB then integrates those equations to compute the kinematics of every body segment and to calculate the forces at all joints. The software can be requested to produce time histories of forces and accelerations of body segments which are used to predict injuries. In this study, the simulations were performed using an interactive version of the ATB, called DYNAMAN [12]. Both DYNAMAN and ATB are inexpensive tools that provide approximate answers to approximate questions.

3.1 Baseline simulation

A detailed description of the simulation of the baseline scenario is given in a previous study [13]. In this baseline simulation, the following features were modeled:

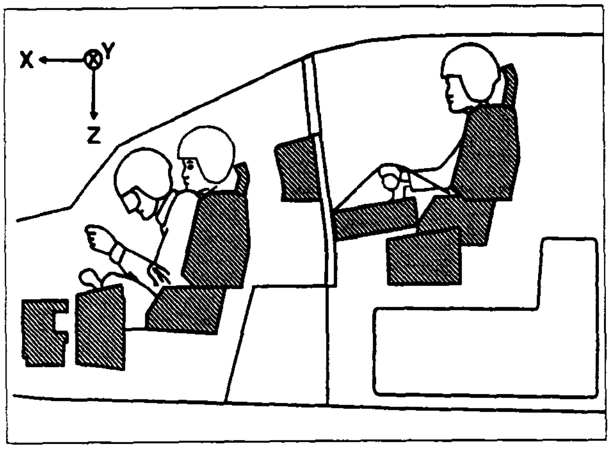

Figure 1 - Seating positions for the pilot (rear) and copilot (front) in the Apache helicopter.

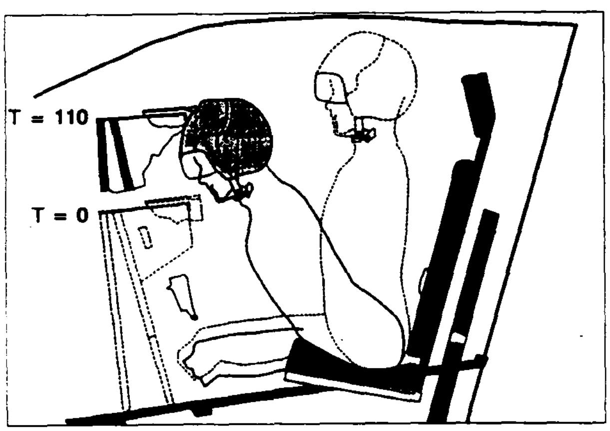

Figure 2 - Likely scenario of a head strike showing pre- and post-crash position of the head relative to the glare shield.

- A vertical acceleration profile to represent the floor acceleration estimated at the rear seat location (Figure 3). The first part of the pulse represents the collapse of the landing gears in the Apache helicopter.

- A mid-size pilot represented by the 50th percentile male Hybrid III manikin data set. The ATB data set for this occupant has been validated extensively.

- A large helmet size simulating the IHADSS helmet worn by the Apache pilots. A small protrusion was modeled to represent the visor knob which is believed to have been caught under the glare shield. The weight of the simulated helmet was 2.7 kg.

- An energy-absorbing seat element that limited the forces transmitted through the seat to the spinal column to no more than 18 kN by stroking. The seat stroke was limited to 2.5 cm travel to simulate the obstruction caused by deformation of the cockpit floor and bulkhead structures.

- A four-point restraint system in which the shoulder harness was not locked until 150 ms into the crash. This simulated a failure of the inertia reel to lock early in the crash because of the low level accelerations during the collapse of the landing gears.

- A rectangular panel was placed in front of the occupant at the same location where the glare shield was mounted in the actual helicopter. Its mechanical properties were estimated from similar panels found in other vehicles.

3.2 Alternate scenarios

Starting from the baseline simulation, alternate "what if" scenarios were simulated. In these scenarios, all parameters were kept as in the baseline except for the following modifications.

3.2.1 Shoulder harness

To explore what would have occurred if the restraint system functioned as expected, the shoulder harness was prevented from excessive extension by locking it at 80 ms into the crash. That is, it was locked when the inertia reel should have sensed the second rise of the pulse after the landing gears had bottomed out.

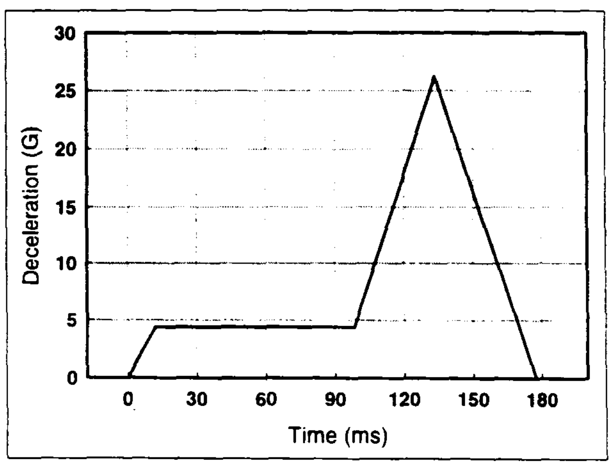

Figure 3 - Vertical deceleration profile estimated at the floor under the rear seat pilot.

3.2.2 Seat stroke
Although stroking of the seat is intended to dissipate energy and limit the forces, it will affect the kinematics of the upper body motion and the excursion of the head during the crash. To explore these effects, the seat was allowed to stroke to 25 cm. The energy-absorbing element in the Apache helicopter seat is designed for a 30-cm stroke.

3.2.3 Airbag model
A multi-airbag system was introduced in the simulations to attenuate the severity of a head strike, when such event occurs. The airbags were represented by ellipsoids with simple geometry and deployment characteristics. The detailed description of the airbags, the panel definition, airbag geometry, deployment history, thermodynamic properties, position and sizes of each airbag were presented elsewhere [14].

The geometry of the airbags and the relative positions of their respective reaction panels, deployment point, and deployment direction are depicted in Figures 4 and 5. Although three airbags (front, left, and right) were available, only the frontal airbag interacted with the pilot because of the nature of the mishap.

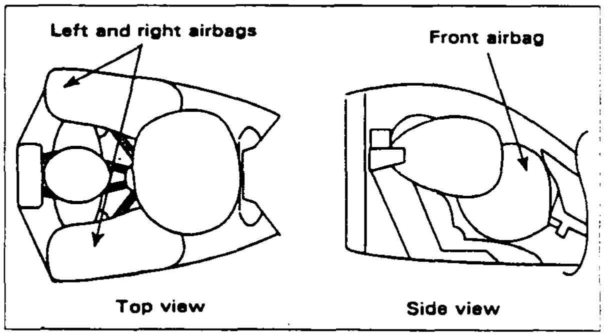

Figure 4 - Top and side views of a simulated multi-airbag system for the Apache helicopter.

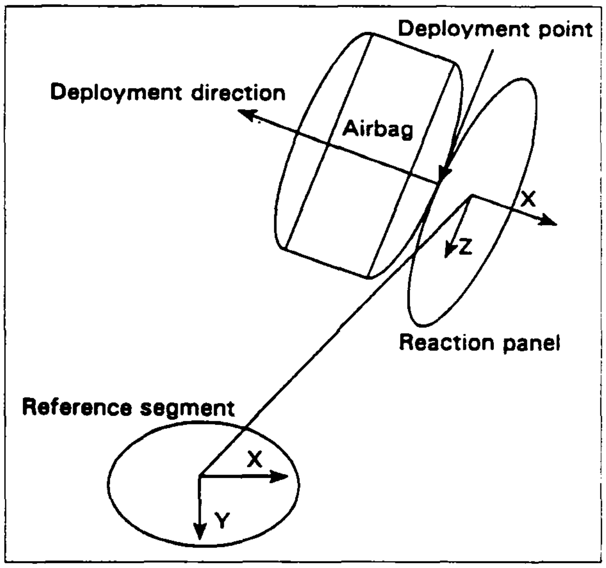

Figure 5 - Geometry of simulated frontal airbag.

3.3 Injury assessment
Injury assessment generally requires the use of crash dummies in actual crash tests to determine if the human body can tolerate the forces of impact. The method of assessment is to compare the magnitudes and durations of individual force and acceleration pulses, measured at strategic locations in the dummy, to acceptable tolerance limits. Given the time history graph of a load, prominent pulses are examined by plotting magnitudes (N, or N·m) within each pulse versus the width of the pulse (ms) at that loading level. Assessment reference values are well defined for the Hybrid III type dummies [15]. In this study, the same assessment methods and reference values were applied to time histories generated by the ATB simulations.

The focus of this study was head and neck injury. In this case, the head injury criterion (HIC) and neck loads at the head-neck interface are commonly used as injury predictors. Since an airbag was used, direct head contacts did not occur in all simulations, making comparisons of HIC values not useful. This leaves neck forces and moments as the only reasonable injury parameters which may be examined. Furthermore, because the simulated motion was primarily in the mid-sagittal plane, the analysis was limited to neck compression and tension forces (±Fz), to fore-aft shear forces (±Fx), and to the extension-flexion moment about the lateral axis (±My).

4. RESULTS
Results of paired simulations are given in Table 1 as peaks of relevant response parameters. These parameters are the compression-tension axial force, the fore-aft shear force and the flexion-extension moment, all computed at the head-neck interface. The listed peaks are exactly as determined from time histories without regard to pulse width. Therefore, one should be cautious in interpreting the differences between the results of different test conditions.

Time histories of response parameters at the head-neck interface are shown in Figures 5, 6, and 7 for simulations in which the harness did not work. Results where the harness locked properly are shown in Figures 8, 9, and 10. In all these simulations, the seat stroke was limited to 2.5 cm travel.

Evaluation of the effects of the locking of the harness (vs. not locking), the seat fully stroking to 25 cm (vs. stroking only 2.5 cm), and the presence of the airbag (vs. not having an airbag) was done by using injury assessment techniques described earlier. It should be noted that this assessment method is used here only as a comparative tool to allow evaluation of results of paired simulations without inferring any injury outcome.

Injury assessment graphs are shown in Figures 11, 12, and 13 for simulations where the harness was allowed to fail by delaying its locking, and in Figures 14, 15, and 16 for the same simulations but with the harness locking as expected. These correspond to the time histories shown in Figures 5 through 10. Again, these results are for runs where the seat stroke was limited to 2.5 cm.

5. DISCUSSION
The peak loads listed in Table 1 are for simulations with 2.5-cm and 25-cm seat strokes. Discussions involving peak loads apply to both seat strokes. However, the discussion of time histories (Figures 5 through 10) and injury assessment (Figures 11 through 16) will be limited in this section to the 2.5-cm seat stroke simulations. We will discuss the results of the baseline simulation, then discuss the effects of the seat stroke, harness failure and the introduction of a frontal airbag on the neck injury response parameters.

Table 1
Peaks of relevant response parameters computed at the head-neck interface.

Response Parameter	Seat stroke (cm)	Harness malfunctions: Without airbag	Harness malfunctions: With airbag	Harness functions: Without airbag	Harness functions: With airbag
Rearward shear force (N)	2.5	8646	912	4308	1744
	25	9020	1495	2011	1015
Forward shear force (N)	2.5	2799	2523	2893	1896
	25	3017	814	2345	841
Compressive force (N)	2.5	7805	1994	7894	2852
	25	6635	2924	3716	1121
Tensile force (N)	2.5	16647	3315	7418	2456
	25	17346	721	3435	1268
Flexion moment (Nm)	2.5	1022	87	588	223
	25	894	215	238	87
Extension moment (Nm)	2.5	658	84	254	146
	25	303	68	324	47

In all these discussions, keep in mind that only the 50th percentile male aviator was simulated. Results will be different for other aviator sizes, particularly for a small size female. It should be noted also that the airbag model used in these simulations is a simple representation of a complex system. Other more sophisticated airbag models have been developed using computational fluid dynamics and finite element methods. These models, which require extensive computational capabilities and resources, were not used in this study.

5.1 Baseline simulation

In the baseline simulations, Beale showed that pilot's helmet became wedged under the glare shield, as displayed in Figure 2. Further forward motion of the neck and body, while the head essentially remained motionless under the glare shield, would have allowed forces of such magnitude, direction, and duration to produce the observed basilar skull fracture.

5.2 Effects of seat stroke

In all simulations where the harness functioned properly, the peak magnitudes of neck loads (shown in Table 1) were reduced by as much as 68% when the seat was allowed to fully stroke. This was true regardless of the presence of an airbag in the simulation. An exception to this observation is the extension moment without airbag which increased slightly when the stroke was increased. In this case, however, the increased moment did not exceed acceptable reference values. The same observations could not be made when the harness malfunctions. That is, no correlation could be found between the seat stroke and peak loads when the harness is simulated to fail.

5.3 Effects of harness

The failure of the inertia reel to lock was simulated by allowing excessive extension. The effects of harness failure can be observed by comparing the injury assessment diagrams of the three response parameters of simulations where the harness worked as expected (Figures 11, 12, and 13) to those where the harness failed to work properly (Figures 14, 15 and 16).

It is clear prominent pulses were reduced significantly when the harness was simulated to work. For cases without airbags, however, prominent pulses in all three response loads remained near borderlines of the assessment corridors, as shown in solid bullets in Figures 14 and 15. When an airbag was introduced along with a working harness, the prominent pulses moved further away from the borderlines and toward the center of the corridors, as shown in the hollow circles of Figures 14 and 15.

Another observation may be made from the time histories of Figures 8, 9, and 10. Even with a working harness but without an airbag, a significant impact with the head rest occurred on rebound (at approximately 325 ms) causing high forces and moments which exceed the corresponding reference values.

5.4 Effects of an airbag

In addition to the airbag-related observations made in the previous paragraphs, two additional observations must be made about the role of the airbag. First, examination of time histories in Figures 5 through 10 shows that all prominent peaks were greatly reduced by the introduction of an airbag. Second, the injury assessments shown in Figures 11, 12 and 13 demonstrate that, even in the absence of a working harness, the airbag reduced magnitudes of prominent pulses to levels that are well within acceptable injury assessment corridors.

6. CONCLUSIONS

In this study, we examined the effects of seat stroke, shoulder restraint and airbags on reducing the severity of potential neck injury to the aircrew during a helicopter crash. For this purpose, we performed mathematical simulations of the pilot's biodynamics to examine the internal loads at the head-neck interface. The simulations demonstrated that, when the harness functioned properly, the peak magnitudes of neck loads were reduced significantly. In the absence of a working harness, results indicated the airbag reduced magnitudes of prominent pulses to levels that are well within acceptable injury assessment corridors.

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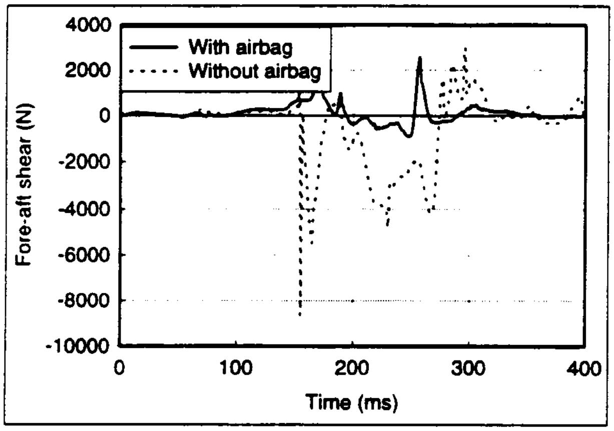

Figure 5 - Fore-aft shear force at the head neck joint in two simulations where the harness did not function as expected.

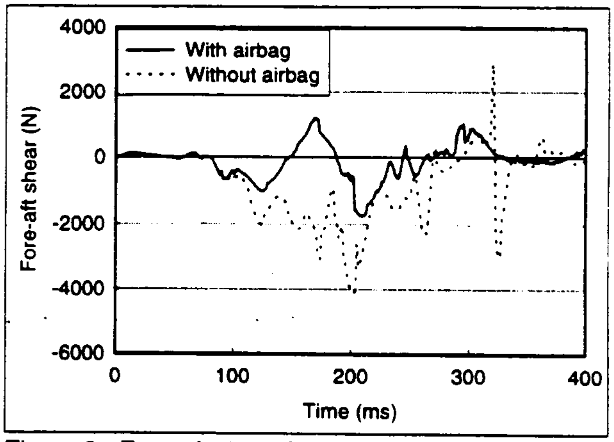

Figure 8 - Fore-aft shear force at the head-neck joint for simulations where the harness worked properly.

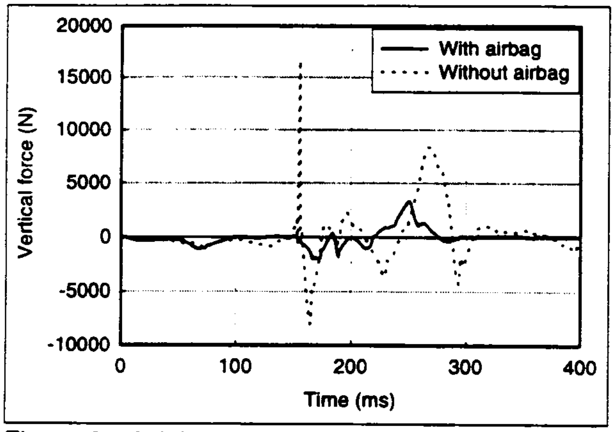

Figure 6 - Axial compression-extension force at the head-neck interface from simulations where the harness did not function properly.

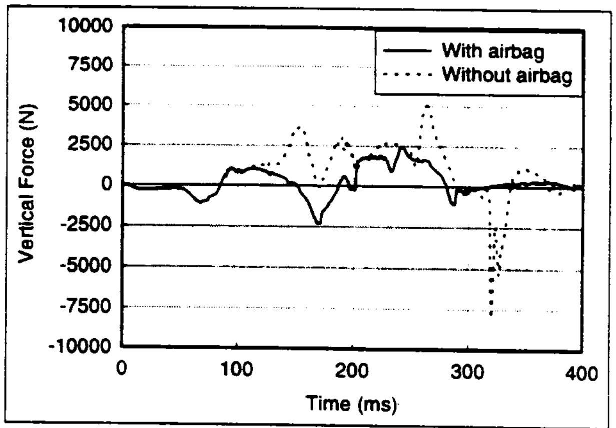

Figure 9 - Compression-extension axial force at the head-neck joint for which the harness worked properly.

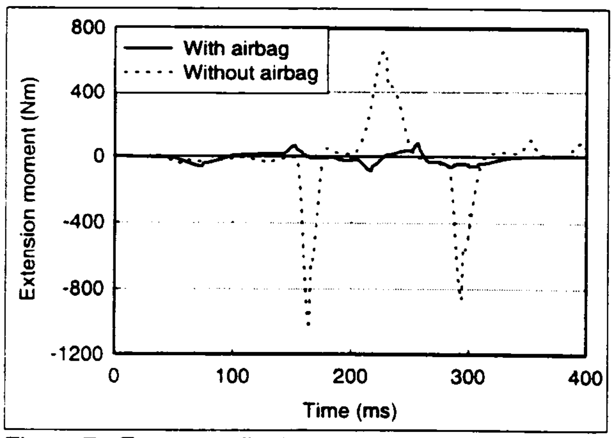

Figure 7 - Extension-flexion moment at the neck joint in two simulations where the harness failed to lock as expected.

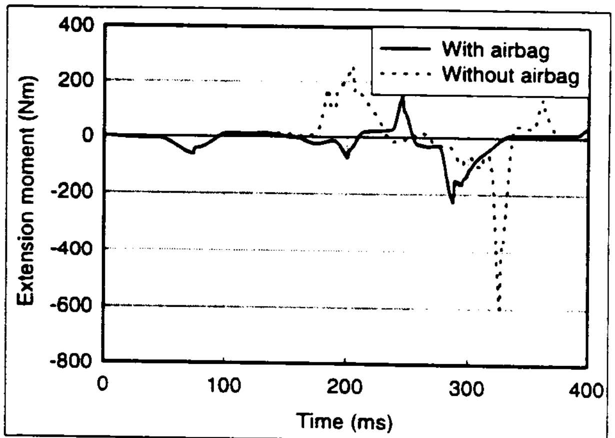

Figure 10. Extension-flexion moment at the head-neck joint for two simulations where the harness functioned as expected.

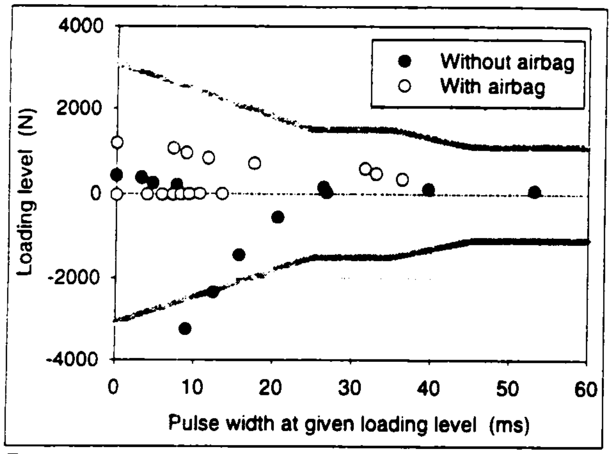

Figure 11 - Injury assessment of neck fore-aft shear from simulated harness malfunction.

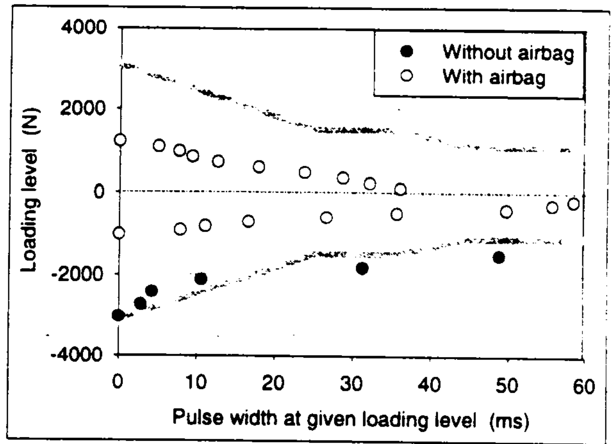

Figure 14 - Injury assessment of neck fore-aft shear from simulations in which the harness functioned properly.

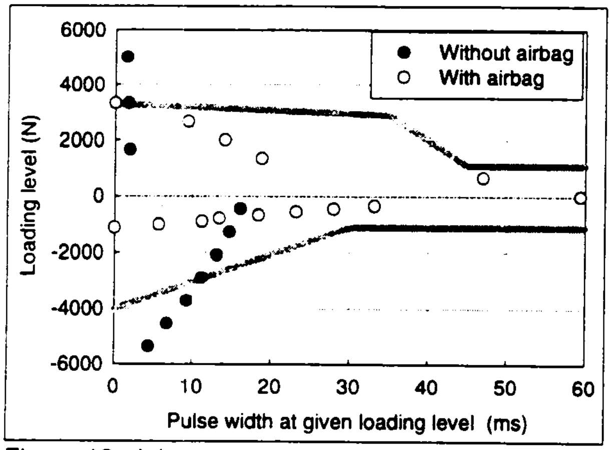

Figure 12 - Injury assessment of neck axial force for simulated malfunctioned harness.

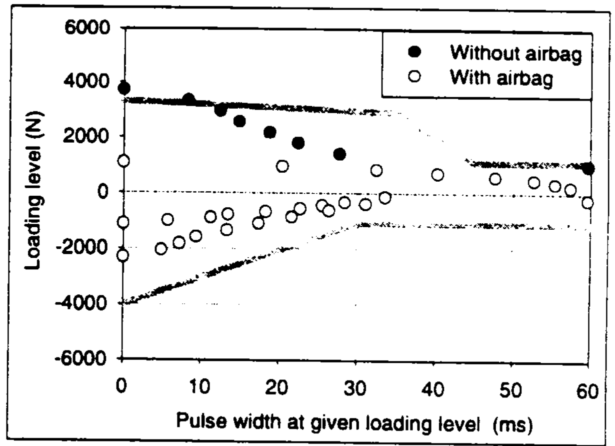

Figure 15 - Injury assessment of neck axial force for simulations where the harness worked properly.

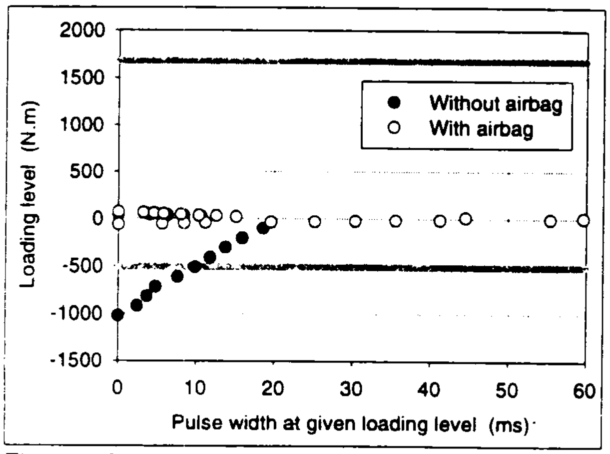

Figure 13 - Injury assessment of neck extension-flexion moment for cases where harness was simulated to malfunction.

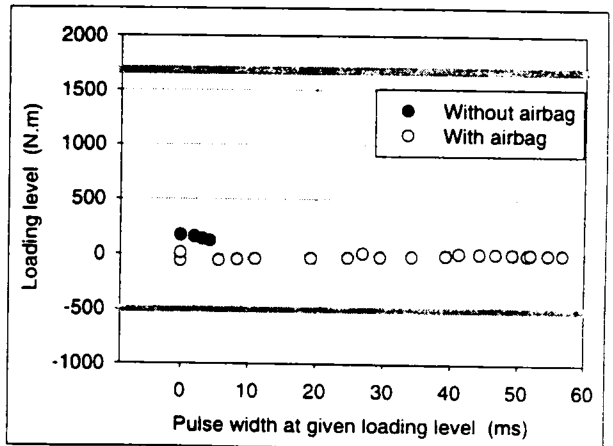

Figure 16 - Injury assessment of neck moment for cases where the shoulder harness was simulated to work properly.